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Sensor and instrumentation for cable tension quantification

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Abstract

A stainless steel based miniature sensor is designed and fabricated for cable tension measurement. The sensor, with a cut out pattern, is only 5 mm high, has an external diameter of 4.5mm and is instrumented with four silicon strain gauges. Applied load will elastically deform the sensor geometry and will induce a resistance change in the Si-piezoresistors. Instrumentation is done using commercial components, and comprise two programmable current sources, that operate the Si-gauges in a current driven Wheatstone bridge configuration. A programmable multi-channel instrumentation amplifier interfaces the Wheatstone bridges with a PIC microcontroller and its on-board 12 bit AD converter. The implemented microcontroller and PC software allows remote adaptation of the analog front and facilitates communication and signal visualization. The sensor system can measure axial forces up to 1000N with a resolution better than 1N.

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*Keywords:*sensor; instrumentation; cable tension;

1. Introduction

Beside the use of bolts, cable based reattachment systems are often used to mount orthopedic implants. F. Canet et al. [1] analyzed the effect of force tightening and simulated walking on cable tension and displacement in trochanter reattachment with an experimental set-up. Dynamic cable load will decrease if the trochanter fracture heals. Initially the stress is transferred to the cable. During fracture healing the forces are gradually transferred to the bone which takes over the function of the cable and the cable becomes obsolete. Image based diagnostic methods to identify the healing status of bone fractures, give limited information due to the ambulatory and large interval data extraction.

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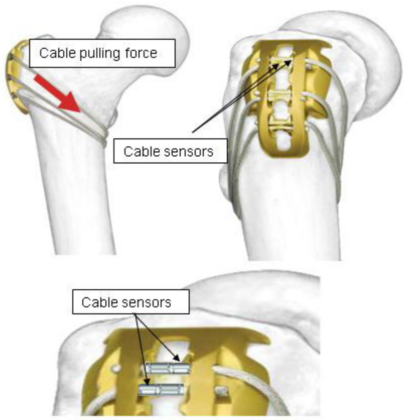

Fig. 1. Cable sensor mounting.

However, the presented novel cable tension sensor, mounted on the orthopedic implant (Fig. 1) and accompanying monitoring circuit could provide patients and orthopedic specialists with in-vivo biomechanical static and dynamic cable measurement data. These allow to identify mechanical behavior, to prevent overloading, follow up greater trochanter healing and adapt prescribed therapy

2. Design, simulation and fabrication

A miniature sensor has been designed and is simulated with COMSOL Multiphysics® software (Fig. 2a), to verify the required dimensions and behaviour of the prescribed cable tension. The designed titanium sensor is 5mm high and has an outside and inside diameter of respectively 4.5mm and 3.8mm. For an applied axial force of 356N the target sensor will deliver 1130µStrain. Its sensitivity is 3.16µS/N. To validate the feasibility of the design, a scaled version of the sensor was manufactured to ease production and testability. By scaling the outside and inside diameter to 6.4mm and 5mm, the sensitivity becomes 3.77 times less or 0.839µS/N for a stainless steel sensor.

The fabricated sensor is constructed out of a stainless steel cylinder by mechanical machining and laser processing (Fig. 2b). 4 SN4-1000-3.8-p-2 Si-gauges from BCM Sensor Technologies are mounted with an cyanoacrylate based Loctite® 460 glue. One reference gauge is glued on one of the three stress free zones. Three measurement gauges positioned 120° apart measure 3 sensor stress components. The Si-gauge golden bond wires are glued with EPO-TEK® H31D, a single component electrically conductive silver epoxy, to the small glued bondable copper terminals (Fig. 2c). The interconnection wires between the sensor and the instrumentation are soldered on the small copper paths to avoid stress transfer to the Si-gauges. An Araldite® two-component glue is used to shield gauges and interconnections.

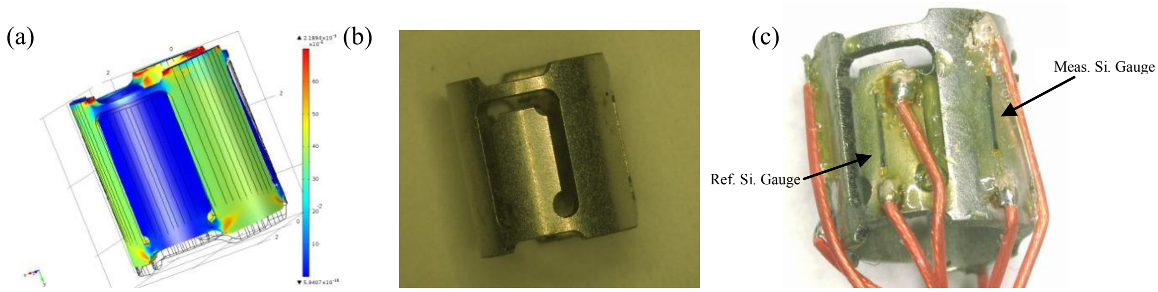

Fig. 2. (a) Cable sensor FEM simulation (b) Realized sensor (C) Sensor with mounted Si-gauges.

3. Instrumentation

An analog front end was designed with low cost commercial off-the-shelf (COTS) components and the functional block diagram is shown in figure 3a. Programmable current sources, based upon the Analog Devices® CN0099 circuit note [2] are used to realize a current driven Wheatstone bridge configuration to compensate for component tolerances and mechanical pre-strain [3]. The current sources exist out of a 2.048V ADR440 precision voltage reference, two AD5063 16 bit digital to analog converters and voltage to current circuits. These are built around the AD8277 precision differential amplifier and the zero drift AD8572 low offset operational amplifier. The full scale current is about 1.5mA and the global maximum current source error is better than 0.15%. The current source resolution is better than 100nA.

One current source drives a 1k Ohm 0.1% precision reference resistor and the selected Si gauge is driven by the second source via a low Ron ADG712 switch. The Si-gauge piezoresistive behavior, which changes resistance and thus the differential bridge voltage $V_{i,sg}$-V_{ref}, selected by a second switch, is amplified by a low offset AD8556 programmable instrumentation amplifier with build in EMI filter. A Microchip® 16F1783 PIC microcontroller with an on-board 12 bit AD converter outputs the measurements and allows remote control of the programmable analog front-end (AFE) via a PC interface. The realized instrumentation PCB 20mm x 33mm large is shown in Fig. 3b.

(a) (b)

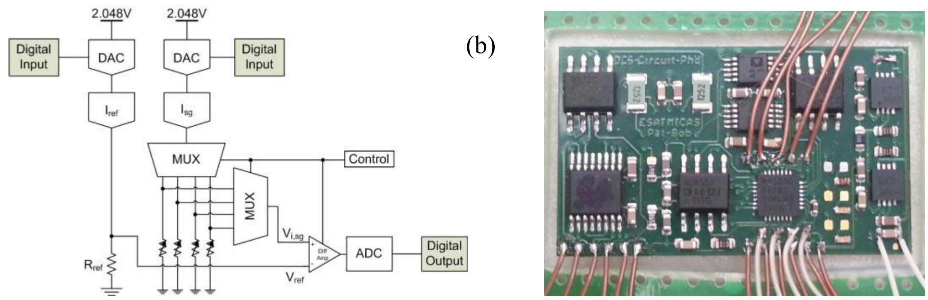

Fig. 3. (a) Current driven Wheatstone AFE. (b) Instrumentation PCB.

4. Measurement results

The sensor has been tested with a pneumatic driven compression tester where the applied force was measured by a 1% precise 1000N S-beam force sensor. Figure 4a illustrates measurement results for 500µA bridge excitation and an amplification of 100. For a load change of 160N (450N .. 290N interval) the measured difference in output signal is approximately 10mV.

(a) (b)

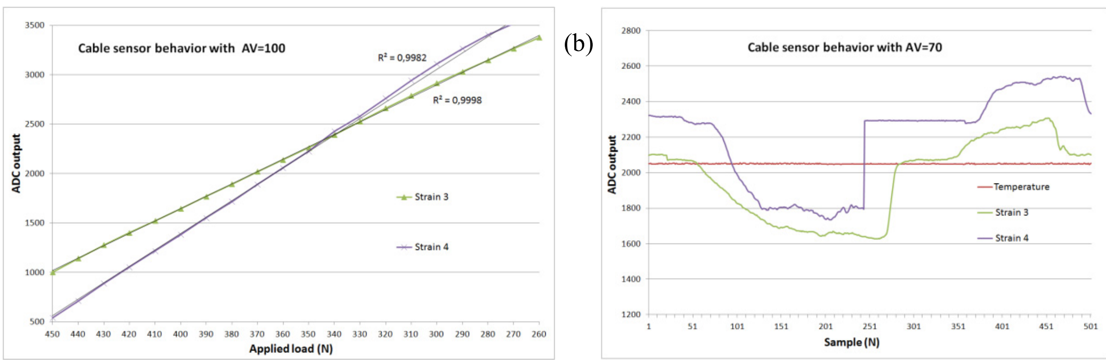

Fig.4. (a) Sensor static load measurement for [250N..450N] applied axial load. (b) Dynamic load [350N..400N] and step response [350N..400N].

This corresponds with a 20 Ohm change and a delta of 133µStrain for the 1k Ohm Si gauge with a gauge factor of 150. So the sensor has a sensitivity of 0.831 µS/N or 62.5 µV/N. For a load of 1000N the strain will be 831µS and is within 1% of the simulated 839µS.

Without temperature compensation, the measurement precision is limited by the Si-gauge resistance and gauge factor temperature coefficients respectively TCR and TCGF. It is possible to calibrate the sensor by measuring the sensor at different temperatures and use the 4^{th} Si gauge as temperature sensor for selecting and/or interpolating between the calibration data. Dynamic sensor behaviour as well as load step response is illustrated in figure 4b sampled at 25Hz during 20 seconds with a gain of 70.

5. Conclusion

Initial measurements done for axial loads between 250N and 450N show a 0.831 µS/N or 62.5 µV/N sensitivity and resolution better than 0.1N. Without temperature compensation, the measurement precision is limited by the Si-gauge 0.30%/°C temperature coefficient. Dynamic behaviour as well as load step response is illustrated.

Further work will focus on improvement of the sensor temperature behavior and improvement of electronic circuitry as well as solving remaining manufacturing, scaling and packaging problems. All of these with special focus on biocompatibility requirements for implantable instrumentation systems. By designing an application specific integrated chip the instrumentation and acquisition could be integrated in the sensor itself.

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